Effect of Capacitor Size and Pathway Resistance on Defibrillation Threshold for Implantable Defibrillators

Charles D. Swerdlow, MD; Robert M. Kass, MD; Peng-Sheng Chen, MD; Chun Hwang, MD; Sharro Raissi, MD

Background The time constant of truncated exponential pulses used with implantable defibrillators is the product of the resistance of the defibrillation pathway and the capacitance of the output capacitor. It is the pulse duration required for the voltage to fall to \( e^{-1} \) (37%) of the initial voltage and to deliver 86% of the stored energy. Because stored energy is an important determinant of pulse generator size, optimizing the size of the output capacitor may permit use of smaller pulse generators. This in turn may simplify pectoral implantation and decrease the morbidity and cost of implantable defibrillator therapy.

Theoretical and empirical models suggest that bio-stimulation can be achieved with minimum energy when the time constant of the stimulating pulse is close to the biologic time constant of the heart. It has been hypothesized that the same principles apply to defibrillating pulses, and most studies of the relation between defibrillation threshold (DFT) and pulse duration confirm that pulses that are either too short or too long require higher energies to defibrillate. A theoretical model predicts that optimal capacitance for monophasic defibrillating pulses is inversely related to pathway resistance if pulse duration is also optimized. However, optimal pulse duration is not known.

To investigate the relation between optimal capacitance and pathway resistance, we compared the DFT for standard 120-\( \mu \)F capacitors (DFT120) and the DFT for smaller 60-\( \mu \)F capacitors (DFT60) at implantation of cardioverter-defibrillators. Multiple electrode configurations were studied to permit observations over a wide range of pathway resistances using clinical defibrillation pathways.

Methods

Patients

Table 1 is a summary of the patients' clinical characteristics. They were studied at the time of new implantation using transvenous electrodes (52 patients) or pulse generator change using epicardial electrodes (15 patients). Patients gave written, informed consent according to a protocol approved by the Human Subjects Committee.

DFT Testing

The DFT was measured after 10 seconds of induced ventricular fibrillation. Truncated exponential pulses with 65% tilt were delivered using an external cardioverter-defibrillator with modular output capacitors (Model 2394, Medtronic Inc). The value of the output capacitor was selected to be either 60 or 120 \( \mu \)F. These values were measured by calculating the tilt of fixed-duration pulses delivered into a calibrated external resistor and taking into account the internal resistance of the...
TABLE 1. Clinical Characteristics of Study Patients (n=67)

<table>
<thead>
<tr>
<th>Characteristic</th>
<th>Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Age, mean±SD, y</td>
<td>64±13</td>
</tr>
<tr>
<td>Male/female (%)</td>
<td>51/16 (76/24)</td>
</tr>
<tr>
<td>Cardiac disease, n (%)</td>
<td></td>
</tr>
<tr>
<td>Coronary artery disease</td>
<td>54 (61)</td>
</tr>
<tr>
<td>Myocardial or valvular disease</td>
<td>12 (18)</td>
</tr>
<tr>
<td>Primary electrical disease</td>
<td>1 (1)</td>
</tr>
<tr>
<td>Clinical arrhythmia, n (%)</td>
<td></td>
</tr>
<tr>
<td>Sustained ventricular tachycardia</td>
<td>42 (63)</td>
</tr>
<tr>
<td>Ventricular fibrillation</td>
<td>18 (27)</td>
</tr>
<tr>
<td>Both</td>
<td>7 (10)</td>
</tr>
<tr>
<td>Left ventricular ejection fraction (mean±SD)</td>
<td>0.32±0.12</td>
</tr>
<tr>
<td>Prior cardiac surgery, n (%)</td>
<td>35 (52)</td>
</tr>
</tbody>
</table>

defibrillator. Based on a mean of 10 determinations, these values of capacitance were 121.3±2.0 and 61.9±1.0 μF, respectively. We tested both monophasic pulses and asymmetric biphasic pulses with 0.2-millisecond separation between the phases.2,12,13 If the defibrillation test pulse was unsuccessful, immediate external rescue was performed by a transthoracic defibrillator. The minimum rest period after defibrillation was 3 minutes. Fibrillation was not induced again until the ST segments and arterial pressure had returned to baseline.

The programmed energy of the initial test pulse was 10 J for epicardial shocks and 15 J for transvenous shocks. It was increased if the test shock failed and decreased if the test shock succeeded. Programmed increments were 5 J between energies of 25 and 15 J, 2.5 J between energies of 15 and 2.5 J, and 1.25 J between energies of 2.5 and 1.25 J. The DFT was defined as the lowest measured energy that terminated ventricular fibrillation. The maximum energy of 25 J was selected because it corresponded to 910 V for 60-μF capacitors, and the safety of defibrillation with higher voltages has not been investigated extensively. For pathways in which the DFT exceeded 25 J for 120-μF shocks, the value of 30 J was used for data analysis.

In each patient, we performed a paired comparison of 60- and 120-μF capacitors in random order. The first 3 epicardial patients and first 25 transvenous patients were tested using either monophasic or biphasic pulses. None of these patients suffered adverse consequences, so the remaining patients were tested using both monophasic and biphasic pulses if the investigator judged that their clinical condition permitted additional DFT determinations. When both monophasic and biphasic pulses were tested in a single patient, their order was also selected randomly. First, both sizes of capacitors were tested for the first waveform; then, both sizes were tested for the second waveform. In 60 patients, only a single electrode configuration was tested. In 3 patients, we tested two electrode configurations using biphasic pulses; the order of electrode configurations was selected randomly. After each shock, patients were monitored for ST-segment depression, hypotension, and transient heart block.

Defibrillation Configurations

Epicardial pathways used a large (conductance area, 840 mm²) patch electrode positioned posteriorly over the left ventricle as the cathode and a large (7 patients) or medium (conductance area, 660 mm²) patch electrode (8 patients) positioned anteriorly over the right ventricle as the anode. Four different clinically used, transvenous, implantable electrode systems were tested.11,14,16 Two used only transvenous electrodes, and two included subcutaneous or submuscular electrodes. In 9 patients, we used the ENDOTAK model 64 electrode (Cardiac Pacemakers Inc). The distal coil (conductive area, 295 mm²) in the right ventricle served as the cathode for the first phase of biphasic shocks; the proximal coil (conductive area, 617 mm²) located at the junction of the right atrium and superior vena cava (low superior vena cava position) served as the anode. In 8 patients, we used an electrode in the right ventricle (Model 6986 or 6986, Medtronic; conductive area, 426 mm²) as the cathode for the first phase of biphasic shocks. A unipolar electrode with a 5-cm-long defibrillation coil (Model 6963, Medtronic; conductive area, 90 mm²) served as the anode. It was positioned via the left subclavian vein with the tip at the junction of the innominate vein and superior vena cava (high superior vena cava position). In 24 patients, we used the same Medtronic right ventricular electrode as the cathode for the first phase of biphasic shocks; a subcutaneous patch electrode (Model 6999, Medtronic; conductive area, 660 mm²) positioned retropectorally in the left infracavicular region served as the anode. Seventeen of these 24 patients were tested using monophasic shocks with the right ventricular electrode as the cathode. In 12 patients, we used the same Medtronic electrodes to combine a transvenous-subcutaneous pathway and a second transvenous-transvenous pathway in parallel for monophasic shocks. The right ventricular electrode served as the common cathode. The patch electrode positioned prepectorally in the left infracavicular region and the unipolar electrode in the posterolateral coronary sinus were linked to form the anode.

Data Acquisition

In all patients, voltage and current waveforms were recorded on a digital oscilloscope (Model 2230, Tektronix) and measured with electronic calipers. In the last 29 patients, voltage and current waveforms were digitized at 100 kHz (MacAdios Board; GW Instruments) and recorded on a Macintosh computer. A custom-modified oscilloscope emulation program (SUPERSCOPE II; GW Instruments) was used to detect leading- and trailing-edge voltages and currents. Automated measurements were confirmed with digital calipers at high resolution on the computer screen; measurements on the oscilloscope were used for confirmation. The maximum difference between the two methods was 2%. Pathway resistance was determined as the ratio of the leading-edge voltage to the leading-edge current. The pulse width of biphasic pulses was reported as the sum of the pulse widths of each phase, excluding the 0.2-millisecond delay between phases. Stored energy was calculated from the measured value of the capacitors, the leading-edge voltage, and the known internal resistance of the defibrillator.12,17 Delivered energy was calculated as the difference between the initial and final stored energies.2

Statistical Analysis

The relation between pathway resistance and the ratio of DFT60 to DFT120 was assessed by linear and polynomial regression using polynomials up to the fourth order. Additional effects of waveform (biphasic or monophasic) and electrode configuration on the ratio of DFT60 to DFT120 were assessed by two-factor ANCOVA using electrode configuration and waveform as factors, pathway resistance as the covariate, and the ratio of DFT60 to DFT120 as the independent variable. For illustrative purposes, DFT60 and DFT120 were also compared for different electrode configurations and pathways with different resistances by the paired t test. Possible differences in clinical characteristics among patients who were tested with different electrode configurations were assessed by ANOVA using clinical variables listed in Table 1 as independent variables. A value of P<0.05 was considered significant. When multiple t tests were performed, a value of P=.05 divided by the number of comparisons was considered significant.18

Results

Defibrillation Waveforms

DFT60 and DFT120 were compared for 44 pathways (44 patients) using monophasic pulses and 53 pathways
(50 patients) using biphasic pulses. Fig 1 shows recorded voltage waveforms for programmed 10-J shocks delivered between a right ventricular cathode and left retropectoral anode. The voltages and currents are higher and the pulse width is shorter for the 60-μF capacitor. The width of the second phase of each pulse is slightly greater than the width of the first phase because resistance is higher for the lower-voltage second phase.19,20

**DFTs**

Table 2 is a summary of DFT data for all pathways. The voltage and current DFTs were substantially higher for 60-μF capacitors, and the pathway resistance at the DFT was slightly lower.

**Ratio of DFT60 to DFT120 Versus Pathway Resistance**

Fig 2 shows the relation between the ratio of DFT60 to DFT120 and pathway resistance. Each data point represents a paired comparison in a single pathway. When pooled data from all electrode configurations are analyzed, there is a significant inverse correlation ($P<.0001$) for both monophasic pulses and biphasic pulses: 120-μF capacitors are superior for low-resistance pathways, and 60-μF capacitors are superior for high-resistance pathways. The linear correlation is slightly stronger for monophasic pulses than for biphasic pulses ($r=.75$ versus $r=.68$), and the slope of the regression line is steeper ($-.19$ versus $-.14$), but these differences were not significant. Higher-order polynomial functions had weak higher-order terms and did not improve the correlation.

Although there is considerable scatter in both the monophasic and biphasic plots, the points for all electrode configurations lie generally along a continuum. Subgroup analysis for each electrode configuration is summarized in Table 3. Despite the narrow range of pathway resistances and smaller sample size for each electrode configuration, there is a significant inverse relation within two of three configurations tested using monophasic waveforms and two of four configurations tested using biphasic waveforms.

Two-factor ANCOVA demonstrated a significant effect on the ratio of DFT60 to DFT120 only for pathway

<table>
<thead>
<tr>
<th>Table 2. Effect of Capacitor Size on Defibrillation Threshold</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Epicardial Pathways</strong></td>
</tr>
<tr>
<td><strong>Monophasic</strong> (n=15)</td>
</tr>
<tr>
<td>Stored energy, J</td>
</tr>
<tr>
<td>Delivered energy, J</td>
</tr>
<tr>
<td>Voltage, V</td>
</tr>
<tr>
<td>Current, A</td>
</tr>
<tr>
<td>Resistance, Ω</td>
</tr>
<tr>
<td>Pulse width, ms</td>
</tr>
</tbody>
</table>

Fig 1. Voltage waveforms (top) and current waveforms (bottom) for 10-J asymmetrical biphasic pulses delivered from right ventricle to submuscular patch. Waveforms for 120-μF capacitors are shown on the left, and waveforms for 60-μF capacitors are shown on the right. Waveforms were digitized at 100 kHz and recorded on a Macintosh computer. The tilt of each phase is 65%. The trailing-edge voltage of the first phase equals the leading-edge voltage of the second phase. The phases are separated by 0.2 millisecond. The programmed leading-edge voltage was 410 V for the 120-μF capacitor and 580 V for the 60-μF capacitor. The measured voltages were 420 and 611 V, and the corresponding calculated stored energies are 10.7 and 11.5 J, respectively. The measured leading-edge current is 5.21 A for the 120-μF capacitor and 6.88 A for the 60-μF capacitor; the corresponding calculated pathway resistances are 81 and 70 Ω, respectively. When monophasic pulses were used, only the first phase of each pulse was delivered.
TABLE 3. Data Stratified by Electrode Configuration

<table>
<thead>
<tr>
<th>Electrode Configuration</th>
<th>n</th>
<th>P</th>
<th>r</th>
<th>Resistance, mean±SD, Ω*</th>
<th>DFT120, J</th>
<th>DFT60, J</th>
<th>P</th>
</tr>
</thead>
<tbody>
<tr>
<td>Monophasic pulses</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Epicardial</td>
<td>15</td>
<td>&lt;.02</td>
<td>.60</td>
<td>39±7</td>
<td>6.5±4.9</td>
<td>7.4±5.0</td>
<td>.002</td>
</tr>
<tr>
<td>RV→CS+subcutaneous</td>
<td>12</td>
<td>(.27)</td>
<td>.35</td>
<td>53±11</td>
<td>15.9±6.6</td>
<td>13.0±4.9</td>
<td>NS</td>
</tr>
<tr>
<td>RV→submuscular</td>
<td>17</td>
<td>.0008</td>
<td>.73</td>
<td>72±11</td>
<td>22.0±6.8</td>
<td>13.7±5.6</td>
<td>.001</td>
</tr>
<tr>
<td>Biphasic pulses</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Epicardial</td>
<td>12</td>
<td>(.21)</td>
<td>.38</td>
<td>39±7</td>
<td>4.6±3.5</td>
<td>5.3±3.6</td>
<td>.05</td>
</tr>
<tr>
<td>RV→low SVC</td>
<td>9</td>
<td>NS</td>
<td>.09</td>
<td>52±4</td>
<td>11.1±5.5</td>
<td>10.0±5.3</td>
<td>NS</td>
</tr>
<tr>
<td>RV→high SVC</td>
<td>8</td>
<td>.03</td>
<td>.75</td>
<td>59±19</td>
<td>15.3±6.8</td>
<td>14.7±5.1</td>
<td>NS</td>
</tr>
<tr>
<td>RV→submuscular</td>
<td>24</td>
<td>.0002</td>
<td>.70</td>
<td>73±12</td>
<td>11.9±5.5</td>
<td>9.0±5.4</td>
<td>.004</td>
</tr>
</tbody>
</table>

DFT indicates defibrillation threshold; RV, right ventricle; CS, coronary sinus; and SVC, superior vena cava.

*Pathway resistance for 120-μF capacitors at the DFT.

Clinical Implications

The histograms in Fig 3 show DFTs stratified by pathway resistance. For pathways with resistance ≤40 Ω, the modest advantage of 120-μF capacitors is of limited clinical significance because it applies primarily to pathways with low DFTs: 8.2±6.1 J versus 9.6±5.4 J (P<.001) for monophasic pulses and 4.1±2.8 versus 5.1±3.1 J (P<.02) for biphasic pulses. In contrast, the greatest advantage of 60-μF capacitors for pathways with resistance ≥61 Ω is clinically more important because it applies to high-resistance pathways with higher DFTs: 12.4±4.3 J versus 23.1±6.1 J (P<.0001) for monophasic pulses and 8.5±4.9 versus 12.5±6.4 J (P=.0002) for biphasic pulses.

The clinical advantage of 60-μF capacitors for high-impedance pathways is illustrated by the percentile plots of DFTs for pathways with resistance ≥61 Ω shown in Fig 4. For monophasic pulses, the DFT was ≤15 J for 19% of pathways using 120-μF capacitors versus 95% for 60-μF capacitors. Similarly, for biphasic pulses, the DFT was ≤10 J for 48% of pathways using 120-μF capacitors versus 83% for 60-μF capacitors.

Electrode Configurations

There were no significant differences in any of the clinical variables shown in Table 1 when patients were stratified by electrode configuration. Table 3 is a summary of DFT and resistance data for each electrode configuration. For the electrode configuration with lowest pathway resistances (epicardial), 120-μF capacitors had an advantage, although this difference was not significant for biphasic waveforms when corrected for multiple comparisons. For the electrode configuration with highest pathway resistances (right ventricle to submuscular patch), 60-μF capacitors had a significant advantage for both monophasic and biphasic waveforms. Neither capacitor size had an advantage for electrode configurations that had intermediate values of pathway resistance.
was moderately higher for clinical forms. The number of patients in each group and P values are shown in parentheses.

Safety

No patient who completed the protocol developed a fall in arterial pressure of more than 20% or 1 mm of ST-segment depression that persisted for more than 30 seconds. The protocol was aborted in one patient who developed hypotension and responded rapidly to a transient vasopressor infusion. Four patients with severe clinical heart failure and one patient with frequent, spontaneous ventricular tachycardia were observed after surgery in the intensive care unit; the remaining 62 patients returned to monitored beds. All patients ambulated on postoperative day 0 or 1. No patient developed hypotension or exacerbation of heart failure in the perioperative period. Plasma concentrations of creatine kinase–MB fraction were not determined routinely, but the maximum value was normal (2.5 IU) in the one patient who developed inotrope-related hypotension.

Discussion

In the present study, we compared DFTs for conventional 120-μF capacitors and for smaller 60-μF capacitors for clinical electrode configurations and waveforms. The results show an inverse relation between optimal capacitance and pathway resistance: DFT60 was moderately higher than DFT120 for low-resistance pathways, substantially lower than DFT120 for high-resistance pathways, and comparable to DFT120 for intermediate-resistance pathways. This general relation applied to both monophasic and biphasic pulses over a wide range of pathway resistances. Thus, no single value of capacitance will minimize the DFT for all pathways; rather, capacitance should be optimized for the resistance of each pathway.

Comparison With Previous Studies

There are three published (preliminary) reports of the effect of varying capacitance for implantable cardioverter-defibrillators. Comparisons of 85-μF capacitors with 140-μF capacitors using epicardial electrodes in pigs showed a 27% decrease in stored energy DFT for 85-μF capacitors using monophasic pulses and a 21% decrease using biphasic pulses. The mean pathway resistance was 57 Ω in one study and not reported in the other. A third study using biphasic pulses delivered between transvenous and subcutaneous electrodes in humans reported an insignificant 14% decrease in stored energy DFT for 60-μF capacitors compared with 120-μF capacitors. The mean pathway resistance was 60 Ω. To the best of our knowledge, the present study is the first systematic investigation of the relation between optimal capacitance and pathway resistance over a wide range of pathway resistances.

Comparison With Theoretical Predictions

Kroll used the Weiss-Lapicque empiric model of biostimulation and Bourland's average-current ap-
proximation to develop a model of monophasic defibrillation for truncated exponential pulses. This model predicts that optimal capacitance is inversely related to pathway resistance, providing that pulse duration is also optimized. The model also predicts that for fixed-till pulses used in our study, there is a complex inverse relation between the ratio of DFT60 to DFT120 and pathway resistance; the predicted ratio is bounded by 2.0 for low-resistance pathways and 0.5 for high-resistance pathways (see “Appendix”). In the present study, we confirmed this predicted inverse relation qualitatively, and most data points fell within the predicted range. The absence of a more accurate curve fit may be due to limitations of the single-point method we used for measuring DFT and/or limitations of the average-current approximation. No comparable model of biphasic defibrillation has been proposed.

Voltage Dependence of Resistance

Current is considered to be a better physiological measure of DFT than energy, but stored energy is a more direct determinant of implanted pulse generator size. Lower values of capacitance may decrease the stored energy DFT by a small, direct effect on pathway resistance that is independent of the current DFT: the resistance of defibrillation pathways varies inversely with applied voltage. If constant energy pulses are delivered from capacitors of different size, the leading-edge currents will differ by more than the ratio of the leading-edge voltages, since the pathway resistance will be lower for the smaller capacitor operating at the higher voltage.

Study Limitations

The principal limitation of this study is that we pooled data over all electrode configurations to achieve a wide range of pathway resistances in the analysis of the ratio of DFT60 to DFT120 versus pathway resistance. We cannot exclude possible effects of electrode configurations on the ratio of DFT60 to DFT120. However, the predominant effect of pathway resistance was supported by ANCOVA. Furthermore, subgroup analysis of four of the seven combinations of electrode configurations and waveforms confirmed this relation despite their narrower range of pathway resistances and smaller sample sizes.

A second important limitation is that our findings apply specifically to the comparison of 60- and 120-μF capacitors using the waveforms tested.

We consider three other limitations to be less significant: (1) the single-point method we used for measuring DFT is less accurate than other methods but is commonly accepted in human studies because of clinical limitations on repeated episodes of ventricular fibrillation; (2) epicardial pathways were studied with chronically implanted electrodes, and transvenous pathways were studied acutely; and (3) by estimating the DFT at 30 J whenever it exceeded 25 J, we may have underestimated the advantage of 60-μF capacitors for high-resistance pathways. However, this would only strengthen our conclusion regarding the advantage of 60-μF capacitors for these pathways.

Clinical Implications

The size of implantable pulse generators depends only weakly on capacitor size, but it depends strongly on the requirement for stored energy and therefore on the stored energy DFT. Output capacitors in present implantable cardioverter-defibrillators vary from 120 to 150 F.

The size of output capacitors is constrained by the combined requirements to store sufficient energy to defibrillate reliably and to operate below voltages at which components of the output circuit fail or myocardial damage occurs. The output circuits of current pulse generators can operate safely at voltages over 1000 V. There is substantial reason to believe that voltages higher than 750 V (corresponding to 34 J from a 120-μF capacitor) may be safe. During defibrillator implantation, internal 815-V (40 J) rescue shocks are given routinely, pulses up to 1000 V have been shown to be safe in dogs, and humans have been defibrillated safely with voltages up to 1420 V. In the present study, we detected no adverse effects from 910-V pulses (25 J from 60-μF capacitors). However, additional safety studies are required before voltages greater than 750 V can be recommended for implanted pulse generators.

For high-resistance pathways, our findings show that 60-μF capacitors provide lower DFTs than conventional 120-μF capacitors. They may have even a greater advantage over 150-μF capacitors. Assuming a conservative 10-J safety margin, a 25-J monophasic pulse generator would be sufficient for 95% of these pathways versus 19% for 120-μF capacitors. Similarly, a 20-J biphasic pulse generator would be sufficient for 83% of these pathways versus 48% for 120-μF capacitors. Pulse generators with 25-J and 20-J maximal output should permit size reductions of 16 cm³ and 24 cm³, respectively, from the 83-cm³ smallest 34-J pulse generator. A series of pulse generators with varying capacitor size may permit optimal capacitance for each pathway.

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Appendix

We begin with Equation 3 from Kroll:  

\[ I_{\text{avg}} = I_a \left( 1 + \frac{d}{d} \right) \]  

(1)

where \( I_a \) is average current, \( I \) is the rheobase current, \( d \) is the pulse width, and \( d \) is the chronaxie. The rheobase is the average current for defibrillation at infinite pulse width, and the chronaxie is the pulse duration that requires a doubling of the minimum effective peak current. For a fixed-till truncated exponential pulse, the average current at the defibrillation threshold (DFT) is given by the following expression:

\[ I_{\text{avg}} = \frac{I}{d} \int_{0}^{d} e^{-\frac{t}{d}} dt \]  

(2a)

where \( I \) is the peak current at the DFT, \( R \) is pathway resistance, and \( C \) is capacitance. Integrating, we obtain the following:

\[ I_{\text{avg}} = \frac{I}{d} \left( 1 - e^{-\frac{d}{d}} \right) = \frac{I}{d} \left( \text{Tilt} \right) \]  

(2b)
By equating the expressions for $I_{ew}$ from Equations 1 and 2 and solving for $I_0$, we get the following:

$$I_0 = \frac{I(d+d_0)}{RC(Tilt)}$$

With Ohm's law, we get the following:

$$V_0 = \frac{I(d+d_0)}{C(Tilt)}$$

where $V_0$ is the peak current at the DFT. Since stored energy is given by the expression $(CV)/2$, the stored energy DFT is:

$$DFT = \frac{I(d+d_0)^2}{2C(Tilt)^2}$$

Therefore:

$$DFT_{60} = \frac{2(d_{60\alpha}+d_0)^2}{2(d_{120\alpha}+d_0)^2}$$

Since $d_{120\alpha} = 2d_{60\alpha}$,

$$DFT_{60} = \frac{2(d_{60\alpha}+d_0)^2}{2(2d_{60\alpha}+d_0)^2}$$

Pulse duration is a constant multiple of RC for fixed-tilt pulses. The multiple is 1 for 63% tilt. Using the approximation $RC = d$ for the 65% pulses in this study, we obtain the following:

$$DFT_{60} = \frac{2(60 \times 10^{-6}R+d_0)^2}{2(120 \times 10^{-6}R+d_0)^2}$$

Therefore:

$$\lim_{R \to 0} \left( \frac{DFT_{60}}{DFT_{120}} \right) = 2$$

and

$$\lim_{R \to \infty} \left( \frac{DFT_{60}}{DFT_{120}} \right) = 0.5$$

Thus, the ratio of DFT60 to DFT120 is bounded by 2 for low-resistance pathways and 0.5 for high-resistance pathways. Other values lie in an intermediate range. The precise shape of the curve depends on the value of $d_0$, which is not known in humans.

References


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